<u>In the Specification</u>

After paragraph [00016], insert the following new paragraphs [00016.1] – [00016.7]:

[00016.1] In one embodiment, the X-ray tube operates in the range of 35 to 70 Kilovolts with a pixel size of 12 to 120 square millimeters. The backscatter X-ray detector 17 comprises an X-ray sensitive area of 150 to 1500 square inches and is positioned 5 to 15 inches from the body of the person being examined. In another embodiment, the backscatter detector may be positioned farther from the person being examined with a corresponding increase in detector area. It has been empirically determined that these parameters are approximately optimized at the values of 50 Kilovolts, 40 square millimeters pixel area, 952 square inches X-ray sensitive area, and an 8 inch subject-to-detector distance. These technique factors simultaneously provide a CV (coefficient of variation in the range of 2 to 10 percent and X-ray dose in the range of 1 to 5 microRem.

[00016.2] In another embodiment, the X-ray source is composed of an X-ray tube operating at 50 Kilovolts and 5 milliamps, with an inherent filtration equivalent to 1 millimeter aluminum. The X-ray source itself is located approximately 30 inches from the person being examined. Standard references, for example: "Catalogue of Spectral Data for Diagnostic X-rays" by Birch, Marshall and Ardran, Published by The Hospital Physicists' Association, London, 1979.", provide data on X-ray tube output. From this reference, under the conditions created according to the arrangement of the inventors, the subject receives a radiation dose of approximately three (3) microRem with a corresponding X-ray photon flux of about 11,500 X-rays in each pixel. Each of these 11,500 X-rays per pixel will either (1) pass through the body unaffected, (2) be absorbed in the body by the photoelectric effect, or (3) exit the body by Compton scattering.

[00016.3] The image noise in a well designed X-ray imaging system is determined by Poisson statistics based on the average number of X-rays detected in each image pixel. This results in the coefficient of variation (CV), or the relative image noise, being inversely proportional to the square root of the number of detected X-rays:

$$CV = N^{-1/2} \text{ X } 100\%$$

where CV is the coefficient of variation between pixels and N is the average number of X-rays detected per pixel. This can be most easily calculated for a specific image by, for example, localizing several hundred adjacent image pixels located in an anatomically flat area near the center of the subject's body. The mean and standard deviation are calculated by standard statistical methods. The image CV is then 100% times the standard deviation divided by the mean.

[00016.4] The optimum dose of 1 to 10 microRem is determined as the range that satisfies each of the conflicting requirements of image quality and health risk. (As previously discussed, higher radiation doses pose a non-trivial health risk to persons being exposed. Radiation doses lower than this range are unnecessarily conservative for health protection and provide unacceptably poor image quality.) Based on this optimum subject radiation dose the optimum X-ray flux and size of the scanning X-ray beam can be derived. As can be seen in FIG. 7, the difference in X-ray reflections between carbon (Z=6) and oxygen (Z=8) is about 30 percent at the optimum X-ray energy of 30 Kev. It has been empirically determined that in order to detect explosives and other dangerous objects, the acquired image must be able to resolve reflectance differences of a few percent. As discussed later, it has been empirically determined that the CV

must be in the range of 2 to 10 percent. This, in turn, requires approximately 11,500 X-rays to strike each pixel on the subject being examined, as shown by the following analysis. It has been empirically determined that the preferred embodiment detector area of 952 square inches placed 8 inches from the subject will receive a backscatter of approximately 7% of these 11,500 X-rays, i.e. approximately 800 X-rays per pixel. Detective quantum efficiency of 50% will result in about 400 of the 800 X-rays contributing to the detector output signal. As the previously presented equation (2) shows, 400 X-rays detected in each pixel will result in a coefficient of variation of 5 percent, within the required range for a acceptable image quality and detection capability.

[00016.5] Prior art systems are designed and optimized to operate by detecting X-rays that have been transmitted through the body. Even at high X-ray tube potentials, a very low percentage of X-rays can penetrate through the body. Data on X-ray penetration can be readily found in the literature, for example the previously referenced "Catalogue of Spectral Data…" Data from this reference shows that at 140 KV only about 1.4 percent of incident X-rays penetrate through an average person's body thickness of 20 centimeters and are detectable to form an image. Only about 0.19 percent of incident X-rays penetrate through a large person's body thickness of 30 centimeters, and are detectable to form an image. In marked improvement over the prior art, the present invention enables seven percent (7%) of the X-rays to be detected. The greatly increased X-ray fiux utilization of the present invention backscatter geometry can therefore be seen to provide a reduction of a factor of five to thirty-five in the required radiation dose compared to prior art X-ray transmission type systems, for equal numbers of detected X-rays.

[00016.6] The X-ray potential in prior art systems is selected to optimize the image quality for transmission radiography. This requires high energy X-rays to achieve sufficient penetration through the body, and thus achieve a larger number of detected X-rays. For example the system disclosed in U.S. Pat. Re. No. 28,444 operates with a peak voltage of 150 Kilovolts. From the above referenced "Catalog of Spectral Data...", (extrapolating form FIG. 9) this potential produces X-rays with an average energy of approximately sixty (60) Kiloelectron-volts (Kev). The present invention can operate with an empirically and theoretically determined optimum X-ray tube potential of 50 Kilovolts, resulting in X-rays of approximately thirty (30) KeV.

[00016.7] Standard references can be used to obtain the values for the X-ray interaction with matter at these energies, for example, "Radiation Detection" by Tait, published by The Butterworth Group, London, 1979. The attenuation coefficients of carbon and oxygen can be used to show characteristics of the two energies. At 60 KeV, the attenuation coefficients of carbon and oxygen, are 0.1765 and 0.1945 grams per centimeter squared, respectively, or a difference of about ten percent. At 30 Kev, the attenuation coefficients are 0.2520 and 0.3733, respectively, a difference of forty-eight percent (48%). This difference in X-ray attenuation results in a backscatter image contrast at 30 Kev being approximately a factor of five (5) greater than at 60 Kev. It is known that the dose required to produce an image is inversely proportional to the square of the contrast, for example, see"Basic Concepts of Digital Substraction Angiography" by Kruger and Riederer, published by Hall Medical Publishers, Boston, pg 79. Thus, the X-ray energy selection of the present invention provides higher contrast images

allowing the use of approximately a factor of twenty-five (25) times lower radiation dose than prior art systems.